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This report documents the second year of a 4-year project to develop field prototype 3-D ultrasonic medical imaging systems. During the reporting period, progress was made on both 2-D array transducers and on sequential B-scan systems. The major result was development and successful field test in Bosnia and Germany of a sequential B-scan 3-D ultrasound telemedicine system prototype called MUSTPAC-1 (Medical UltraSound, Three-dimensional and Portable, with Advanced Communications). MUSTPAC-1 uses conventional ultrasound probes, operating at 3.5-5 MHz, to obtain high resolution images at relatively low speed (15 seconds per 3-D scan). In addition, feasibility studies were performed regarding the use of 2-D transducer arrays, with the goal of achieving real time image acquisition rates.

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Richard J. Littlefield 10/31/96

PI - Signature Date

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1. INTRODUCTION

Inexpensive, portable diagnostic imaging systems can play a key role in decreasing battlefield fatalities and reducing the cost of military health care. Ultrasound imaging is a particularly promising modality because it does not use ionizing radiation (unlike X-rays), does not require large heavy equipment (unlike magnetic resonance imaging), and has been shown through long use to be safe and effective when used by highly trained practitioners.

However, in current practice, ultrasound is basically an online two-dimensional (2-D) scanning procedure that produces a sequence of images under interactive hands-on control by the diagnostician. Each image represents a slice through the body at the corresponding ultrasound probe position. These images typically are difficult to interpret, requiring a trained ultrasonographer with years of experience to make more than the simplest diagnoses. This need for an expert interpreter makes it attractive to use ultrasound in a telemedicine setting, sending images from the patient's location to a skilled diagnostician somewhere else. However, with conventional 2-D ultrasound, considerable skill is required even to position the sensor probe, since this must be done interactively as diagnosis progresses. Thus it is problematic to use conventional 2-D ultrasound in a telemedicine setting, due to the need for a highly skilled operator to scan the patient.

Three-dimensional (3-D) ultrasound imaging offers the potential to overcome these difficulties, thus providing a diagnostically valuable, low-cost, real-time imaging modality suitable for operation and use under emergency conditions by non-specialists. Because 3-D volumes show more context than 2-D slices, it becomes easier for users to understand spatial relationships and detect abnormal conditions. Positioning the sensor so as to acquire useful images is also easier with 3-D, because volumetric data can readily be rotated and realigned to good viewing positions, largely independent of the original sensor position. This potentially allows useful 3-D ultrasound data to be taken by an inexperienced operator, then transmitted to and interpreted by a remote expert.

While 3-D ultrasound imaging has been investigated periodically for over 20 years, its adoption into routine use has been hindered by clumsy equipment, long image acquisition times, and the difficulties of visualizing the 3-D clouds of relatively noisy data produced by speckle and directional effects of ultrasound. However, recent advances in transducer array technology, computational hardware speed, and improved image reconstruction and visualization methods now appear sufficient to permit these obstacles to be overcome.

These considerations prompted Battelle to submit to ARPA, in response to solicitation BAA94-14 in early 1994, a proposal titled "Real-Time High Resolution 3-D Ultrasonic Imaging for Physiological Monitoring". This proposal laid out the vision of a three stage effort, roughly 8 years in length, leading to the development of an imaging "bed", roughly 10,000 square centimeters in size, containing an array of high resolution ultrasonic transducers and providing real-time 3-D visualization of many physiological and anatomical structures.

The first stage of this vision, and the focus of the proposal, was a 3-year project to develop a field prototype Advanced IMaging System (AIMS). This prototype would consist of a lightweight, portable ultrasonic imaging system envisioned as containing a 5 cm by 5 cm two-dimensional transducer array, computer hardware and software for real-time 3-D holographic image reconstruction and visualization, and a stereovision headset for 3-D image display. The system was envisioned as being used to rapidly detect foreign objects and bleeding in the body cavity, lungs, or extremities.

The Battelle proposal to develop an AIMS prototype was accepted by ARPA, and the project began in September 1994 with two major components:

- Research and develop advanced sensor technology, in particular, 2-D transducer arrays utilizing computational holographic focusing to acquire 3-D images in real time.

- Research and develop one or more fully functional prototype systems, suitable for clinical and/or field use, to investigate and demonstrate the utility of 3-D ultrasound as a medical imaging tool for use by non-specialist operators.

In early stages of the projects, it was planned that these two components would proceed sequentially, with the prototype system(s) being based on newly developed 2-D array transducers and thus appearing late in the project.

However, as the project progressed, it became apparent that a more productive strategy was to pursue both components in parallel, with prototype systems being based on currently available 1-D array transducer technology. This strategy was adopted in FY95 and has proved very effective.

In FY95, results in each research and development component were as follows:

- Sensor technology: Laboratory research using mechanically scanned simulations of 2-D transducer arrays confirmed that high quality images, fully focused everywhere in the 3-D field of view, could be obtained by using computational holographic focusing techniques in conjunction with large arrays (128x128) of high-frequency (5 MHz)

transducers. However, the supporting electronics and computational requirements to use such large arrays appeared beyond the reach of current technology. These requirements could be met with smaller, lower frequency arrays, such as 32x32 at 1MHz. Tests at 1 MHz did not produce encouraging results when used to image commercially available plastic ultrasound phantoms of solid organs (liver and breast). However, the phantoms were not specifically designed for use at such low frequencies, so it was not clear whether the results accurately reflected the medical utility of 1 MHz ultrasound. Accordingly, it was decided to perform further testing at 1 MHz in FY96, using a different type of phantom and focusing on diagnosis of abdominal blood pooling.

- Prototype systems: a clinically usable 3-D ultrasound system based on "sequential B-scan" technology (mechanical sweep of a conventional ultrasound probe) was developed. This system was displayed at the October 1995 annual meeting of the AUSA (Association of the U.S.Army) in Washington DC, where it was favorably reviewed by many Army personnel. More importantly, the system was placed into use in the clinic of Dr.Christian Macedonia at Madigan Army Medical Center in Tacoma, Washington, for an extended evaluation to occur in FY96.

In FY96, research and development continued in both of these areas.

In brief, the most important product from FY96 was a second-generation prototype system, called MUSTPAC-1 (Medical [or Military] UltraSound, Three-dimensional and Portable, with Advanced Communications), that provides powerful telemedicine capabilities. The MUSTPAC-1 was successfully field-tested in Bosnia and Germany in August 1996.

As a result of the success of MUSTPAC-1, a change within scope was negotiated in late FY96 that adjusted the project's priorities to reduce the level of effort on 2-D array developement, emphasizing instead further development of field-usable systems based on sequential B-scan technology.

Further details on work performed in FY96 is outlined in the following section and described in detail in the three appendices.

2. EXPERIMENTAL METHODS AND RESULTS

Activities in FY96 (October 1995 through September 1996) fell primarily in four technical areas:

- Evaluation of 2-D array transducers operating in simultaneous source-receive mode at an array size and frequency consistent with current fabrication technology (32x32, 1 MHz), as applied to abdominal blood pooling using a custom phantom.

  Details of the laboratory portion of this study are contained in the report provided here as Appendix 2: "Evaluation of 3-D Ultrasound Holographic Imaging at 1 MHz for Detection and Visualization of Abdominal Blood Pooling Using a Mechanically Scanned Single Channel Transducer and Laboratory Phantoms".

  This report was reviewed by several medical consultants (Dr. Richard Satava, Walter Reed Medical Center; Dr. Christian Macedonia, Madigan Army Medical Center; Dr. Stephen Carter, University of Washington; and Dr.Jonathan Ophir, University of Texas).

  The concensus opinion was that the images would be at best marginally useful for diagnoisis, and that higher resolution (larger arrays and higher frequencies) would be required to construct a useful system. Since larger arrays and higher frequencies are not practical at present using simultaneous source-receive mode, this line of research has been tabled.

- Evaluation of computational holographic focusing used in conjunction with 2-D array transducers operating in receive-only mode (and thus capable of being fabricated in larger arrays operating at higher frequency).

  Details of this paper study are provided in Appendix 3. The results are encouraging, and further research along this line is planned.

- Evaluation of the FY95-produced "AIMS prototype", as used in clinical studies by Dr.Christian Macedonia at Madigan Army Medical Center.

  Results from this evaluation have not been formally reported, but instead were incorporated directly into the requirements and system design of the MUSTPAC-1.

- Development and field-testing of a successor prototype 3-D ultrasound telemedicine system, called the MUSTPAC-1. This system preserved the high level architecture of the earlier "AIMS prototype", but replaced virtually every hardware and software component of the system with redesigned state-of-the-art elements.

As delivered, the MUSTPAC-1 consists of an 85-pound backpack unit containing everything necessary to operate in telemedicine mode when connected to any standard network supporting TCP/IP. It includes the following subsystems and capabilities:

- Hitachi's Model 905 portable, battery-powered, 2-D ultrasound machine.
- Battelle's 3-D Paddle electromechanical scanner.
- TeleInViVo™ (Fraunhofer CRCG) volumetric visualization software, customized to meet this project's requirements.
- A "virtual ultrasound probe", consisting of an Immersion Probe™ (Immersion Corporation) modified and interfaced by Battelle and Fraunhofer to provide the look, feel, and effect of a conventional ultrasound probe, when computationally reslicing volumetric data.
- Silicon Graphics Indy™ computer and Presenter™ flat-panel display.
- Drag-and-drop and graphical user interfaces for all functions.

Field testing of MUSTPAC-1 was completed in early September, 1996, yielding the following results (as summarized to DARPA by email on Oct.21, 1996):

> In early July, this project delivered a prototype fully functional 3-D ultrasound telemedicine system called the MUSTPAC-1 (Military UltraSound, Three-dimensional and Portable, with Advanced Communication).
>
> MUSTPAC-1 provides the unique capability that high quality ultrasound scans can be taken at forward locations by an operator with no diagnostic skills, little training, and no online connection to an expert. The scans are then transmitted over any standard digital network to a qualified diagnostician, who interprets the scan using a "virtual ultrasound probe" that simulates a conventional real-time hands-on examination procedure. The virtual probe and corresponding screen displays are very natural to diagnosticians, leading to rapid acceptance and productivity (see below).
>
> In August 1996, the MUSTPAC-1 was deployed to the 212th MASH in Tuzla, Boxnia. A second MUSTPAC (minus some packaging) was

deployed to Landstuhl Regional Medical Center in Germany to serve as a "buddy" and consulting workstation for the Tuzla unit. During the 32-day deployment, approximately 72 scans of 38 patients were taken at Tuzla, with an additional 5 scans of 3 patients taken at Landstuhl. Scans were exchanged between Tuzla, Landstuhl, and Madigan Army Medical Center (Tacoma, WA, USA) using three different telecomm networks: the Tuzla/Landstuhl/Tazar teleradiology net, the Tuzla TACNET, and an Inmarsat satellite link (Landstuhl to Washington DC).

The MUSTPAC system worked perfectly.

Trainup times at Landstuhl were observed to be under 5 minutes for diagnosticians to go from first contact with the equipment to making medical interpretations from the 3D datasets. In Bosnia, most of the scans were taken by people with no ultrasound experience, and less than 20 minutes exposure to the equipment. Image quality in the reconstructed scans was excellent.

Further information on the MUSTPAC-1 is provided in Appendix 1.

3. CONCLUSIONS

Success of the MUSTPAC-1 under field conditions has provided considerable validation that 3-D ultrasound can be an effective telemedicine tool, allowing a nonspecialist operator in the field to obtain high quality scans that can be interpreted by a remote expert.

In fact, development and field-testing of the MUSTPAC-1 amounts to early delivery of one major goal of the project: a prototype portable 3-D ultrasound system usable under field conditions by a nonspecialist.

Much further development remains, however, to refine the MUSTPAC-1 technology into a system suitable for routine use under a wide range of conditions. This work is planned as FY97 and FY98 activities within the project.

In addition, sequential B-scan technology (mechanical sweep of a conventional ultrasound probe) is incapable of providing real-time 3-D imaging over a large field at high resolution. 2-D array transducers and computational focusing techniques still appear to be the best

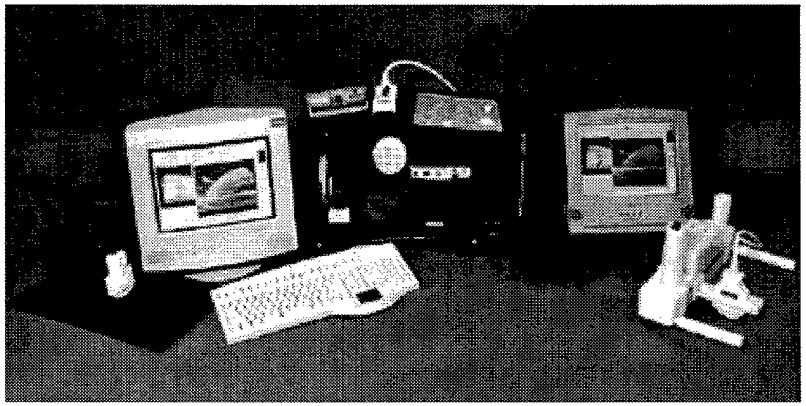
approach to reaching that goal, and these technologies should continue to be researched both within the current project and elsewhere.

Appendix 1:

"MUSTPAC-1: 3-D Ultrasound Telemedicine System"

MUSTPAC-1: 3-D Ultrasound Telemedicine System

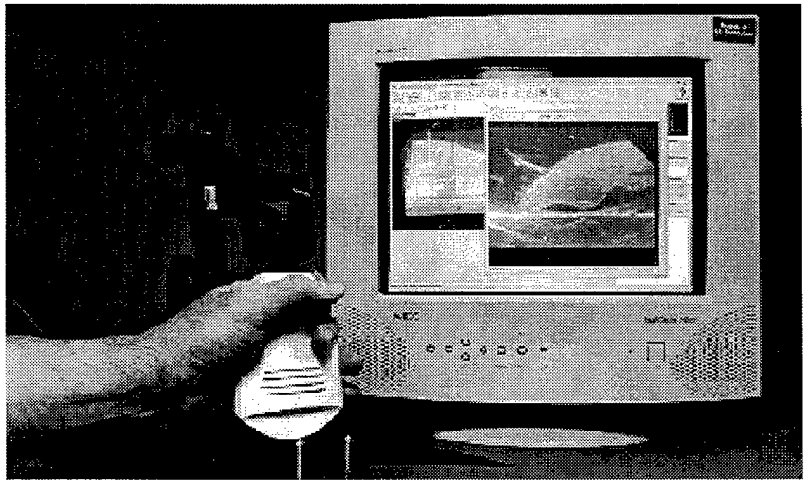
Medical UltraSound, Three-dimensional and Portable with Advanced Communication



The MUSTPAC-1 is a fully functional 3-D ultrasound medical imaging system that incorporates the following elements (**right-to-left** in picture):

- Backpackable field unit containing
  - "3-D Paddle" electromechanical scanner (at extreme right)
  - Silicon Graphics Presenter(TM) flat panel display (to right of backpack)
  - Hitachi EUB-905 ultrasound machine (in backpack, top section, with cord)
  - Silicon Graphics Indy(TM) computer (in backpack bottom section)
  - Teleconferencing camera (on backpack, top left)
  - Keyboard with integral touchpad (in front of backpack)
- High-resolution color monitor.
- Virtual ultrasound probe (Immersion Probe) to provide expert ultrasound diagnosticians with a familiar interface that is easy to learn and easy to use.

The MUSTPAC is unique in that it provides an ultrasound telemedicine capability that is effective, requires very little training for anyone, does not require a highly skilled operator at the patient's side, and operates well even in a low-bandwidth store-and-forward file transfer mode. This is accomplished by 1) scanning a fairly large volume of the patient at one time so that diagnosis during the scan is not required, and 2) providing the remote diagnostician with a familiar interface -- a "virtual ultrasound probe" -- that mimics conventional 2-D ultrasound, so that their years of experience and hand-eye coordination can be brought to bear using the MUSTPAC after only a few minutes practice.

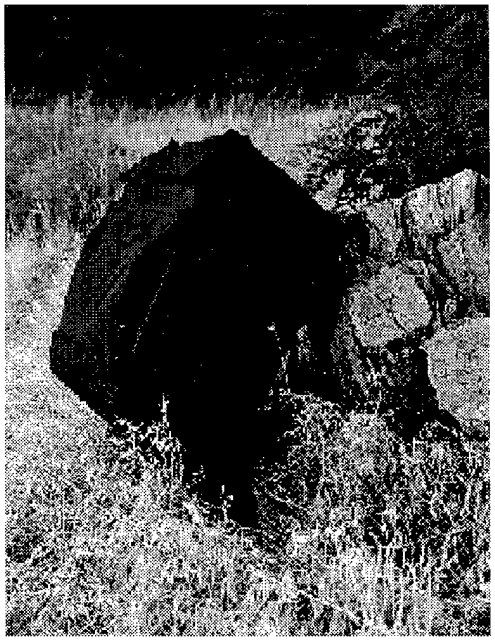


The "virtual ultrasound probe" interface is implemented using a modified Immersion Probe(TM) or equivalent 6-D sensing arm (3-D position, 2-D tilt, plus rotation). Position and orientation of the probe is used by the MUSTPAC's visualization software (TeleInViVo(TM), from Fraunhofer CRCG) to "reslice" the 3-D ultrasound data along arbitrary cutting planes. The screen view is updated in real time (5-10 updates per second) so the diagnosticians can work with MUSTPAC 3-D scans in the same way that they would with a real patient and a conventional 2-D ultrasound system.

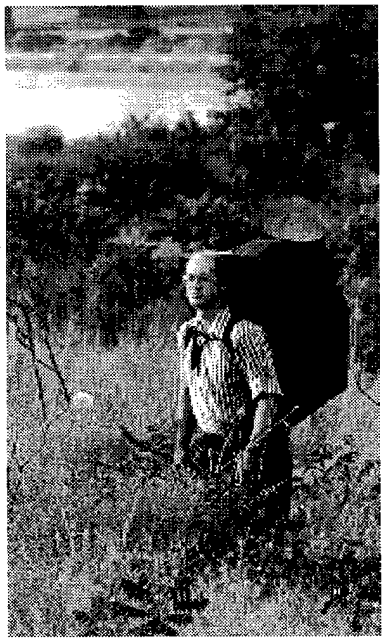
The MUSTPAC-1 prototype is an interim deliverable in a continuing research program funded by DARPA. At present, only one unit has been fully fabricated. However, the system components are all either off-the-shelf or easy to fabricate, so it would be straightforward to produce additional units of the same design.

At present, the MUSTPAC system has only limited approval by the FDA, so that clinical investigations using it require IRB-approved medical protocols with informed consent etc. In experience to date, these approvals have been easy to obtain because the MUSTPAC introduces no significant safety issues -- its sensor technology is just an unmodified conventional 2D ultrasound system, combined with a battery-powered, electrically isolated, mechanical scanner.

Planned future developments include engineering improvements, 510(k) or similar FDA approval for routine use by the military, and clinical studies regarding specific applications.



The MUSTPAC-1 prototype weighs approximately 85 pounds in its backpack configuration. This weight includes an ultrasound system (Hitachi EUB-905), 3D paddle, Indy computer with camera and large LCD screen -- everything needed to acquire 3D ultrasound data, visualize it locally, transmit the data to a remote site, and consult via teleconference. The pack requires AC power and a network connection that supports the TCP/IP protocol. Battery operation has also been demonstrated using a commercial UPS unit (45 minutes, 50 pounds). Satellite operation has been demonstrated using sync routers that transparently extend network operation across a satellite link. Planned system upgrades will result in significantly lower weight, increased battery operation times, and additional communication options.



Development of the MUSTPAC-1 was performed using "rapid prototyping" procedures and based on experience with the predecessor system developed in FY95. Although MUSTPAC-1 had been planned for several months earlier, the final decision to allocate funding and develop a prototype did not occur until April 1996. Acquisition, fabrication, and software development was performed on a very rapid schedule, and the completed backpack was shipped to MATMO at Ft.Detrick on 3 July 96. From there it underwent a series of pre-deployment evaluations under the supervision of the Center for Total Access (CTA, Fort Gordon Georgia). After the performance evaluation was completed, the system was shipped to Landstuhl, Germany, arriving on Friday, 26 July 96. It was unpacked, operational, and running demonstrations within 2 hours of arrival at Landstuhl. Briefings were provided to the Landstuhl commander, COL Kiley, on 30 July 96 and to COL Farmer on 31 July 1996. It was delivered to the European Comand (EUCOM) on 4 August 96.

On 7 August 1996, Dr. Macedonia deployed the MUSTPAC to the 212th Mobile Army Surgical Hospital, Camp Bedrock, Tuzla Bosnia. The machine was operational upon arrival and communications links were established using standard ethernet protocols. Data was transmitted back through a microwave link to Eagle Base Tuzla(12 km away) and then fed through a Ku Band Satilite link to MUSTPAC2 in Landstuhl Germany. Other receiving sites included Richland WA, Tacoma WA, Providence RI, and Washington D.C. The machine also operated in a stand-alone mode. The unit was redeployed to Washington D.C. 8 September 1996.

Appendix 2:

"Evaluation of 3-D Ultrasound Holographic Imaging at 1 MHz for Detection and Visualization of Abdominal Blood Pooling Using a Mechanically Scanned Single Channel Transducer and Laboratory Phantoms"

Appendix A: Evaluation of 3-D Ultrasound Holographic Imaging at 1MHz for Detection and Visualization of Abdominal Blood Pooling Using a Mechanically Scanned Single Channel Transducer and Laboratory Phantoms

(Phase 1, Task 2, report date April 25, 1996)

Task Description

Task 2 is the preliminary experimental evaluation of 1 MHz blood pool images on phantoms, etc., using holographic single channel and "1-D" array systems mechanically scanned at PNNL's Richland facility in the EDL laboratory.

The single channel system will be implemented to experimentally verify the viability of using 1 MHz ultrasound in the detection and imaging of blood pool phantoms before building the first test sub-array at 1 MHz.

The medical trauma community (as reported by Dr. Steven Doctor, PNNL scientist) has successfully used higher frequency (3 to 5 MHz) commercial B-Scanners to detect and image blood pooling in trauma patients. They indicate easy detection at these frequencies using present commercial B-Scanners. Typically, only one day is spent training for blood pooling which indicates it is a simple test with a commercial B-Scanner.

Our task is to evaluate the efficacy of using 1 MHz in the detection and imaging of blood pooling and using this information to construct a "2-D" array capable of "3-D" imaging for a portable combat field system. The decision to use the lower frequency is driven by the complexity and cost of building a high frequency "2-D" transmit / receive array that can be used in the combat "3-D" imaging field system. A simple 32 by 32 array (~5cm by 5cm) at 1 MHz using one wave length element requires 1024 transmit / receive channels for serial switching and many more if parallel and serial switching schemes are used for imaging in real time. The same size array at 3 MHz would require approximately 100 by 100 elements and 10,000 transmit / receive serial switching channels, etc. The higher frequency system would also have greater attenuation and with small transmit elements (with their inability to transmit high power) will decrease depth penetration in blood pool imaging. There are many complex engineering problems to consider when building a high frequency "2-D" transmit / receive array with present day technology. The design and fabrication of a viable "2-D" imaging array has challenged some of the best companies and universities in the United States, Europe and Japan and none exist today in commercial medical equipment.

PNNL will construct a phantom based on information from the medical community, image at 1 MHz using our holographic single channel system, and compare the images with a commercial B-Scanner (3.5 MHz).

The phantom will consist of simple foam (stomach layer) and duct putty (intestines) immersed in water with cavities between the foam and duct putty that simulate blood pools. This assumes the combat casualty is lying on his or her back and the blood rises above the intestines and is trapped by the thick muscle layer (stomach).

Holographic array images (1 MHz) will be integrated into "3-D" volumetric blood pool images for preliminary evaluation of simulated "2-D" array imaging with scanned "1-D" data. These data will simulate the results expected with the "2-D" array that will be used in the battlefield system.

ARPA and expert medical staff will be able to view the single channel phantom blood pool images and determine the viability of using 1 MHz ultrasound for the battlefield "2-D" array blood pool imaging system. If the image resolution (blood pool phantoms) is found insufficient for battlefield diagnostics, PNNL will (with the approval of ARPA) proceed to a higher frequency (~2 MHz), design and fabricate the system. The design will follow along the higher frequency route with the appropriate changes, etc. If the resolution is sufficient at 1 MHz, we will proceed with design and fabrication of the "1-D" array and the "2-D" array combat system.

The small sub-array (few elements) will be designed and fabricated for imaging the blood pool phantom and comparing it with the single channel data. Single channel scanned data, using a one wave length transducer, is the optimum configuration for high resolution imaging and is used as the "gold standard" for comparing array data. The array, with its many elements closely packed together, has various cross-talk and acoustic coupling paths which tend to degrade the image quality as compared with a single channel system. After successful single channel tests at 1 MHz the "1-D", array will be designed and fabricated for testing.

The proposed "1-D", array will consist of 32 elements spaced one wave length apart, approximately 5cm in length. The width of each element is one wave length resulting in a 180 degrees main beam. After design and fabrication of the "1-D" linear array, tests will be conducted using the medical blood pool phantom and the results will be compared with the sub-array data. The results should be comparable if the sub-array data is acceptable for blood pool imaging at 1 MHz.

The "1-D" array when scanned in one direction with the x-y mechanical scanner, (generating many lines of data) will simulate a "2-D" array in non-real time. These tests will be conducted to provide simulated "2-D" array blood pool imaging data for the design of the this array in the next phase. The "2-D" array is the heart of the portable combat real time blood pool imaging system that will be carried by the field medic and used in diagnostic battlefield trauma.

B-Scan (Hitachi) Simulated Blood Pool Imaging of Phantom at 3.5 MHz

A blood pool phantom was constructed of multiple layers of foam (same type as used in the filter of a Tektronix scope fan) and duct putty inserted in a plastic bag (see figure 1). The use of foam was suggested by Dr. Jonathan Ophir, University of Texas medical school who is engaged in research dealing with emergency room trauma using ultrasound as a diagnostic tool for blood pooling. His wife, Karen Ophir, is a registered diagnostic sonographer and also teaches courses in emergency room trauma. Typically, only one day is spent training for detection of blood pooling in trauma patients. The detection is relatively simple using a B-Scan sector array operating above 2 MHz.

A Hitachi portable B-Scan with a 3.5 MHz linear (sector) array was used to image simulated blood pooling using the phantom shown in figure 1. If internal injury severs a large vein or artery in the abdomen, the blood rises (assuming the patient is on his back) upward toward the surface of his stomach and is trapped between the intestines and muscles of the stomach. The blood then forces the upper layer (1.3cm foam) to slightly bulge and form pockets between the intestines and the upper layer.

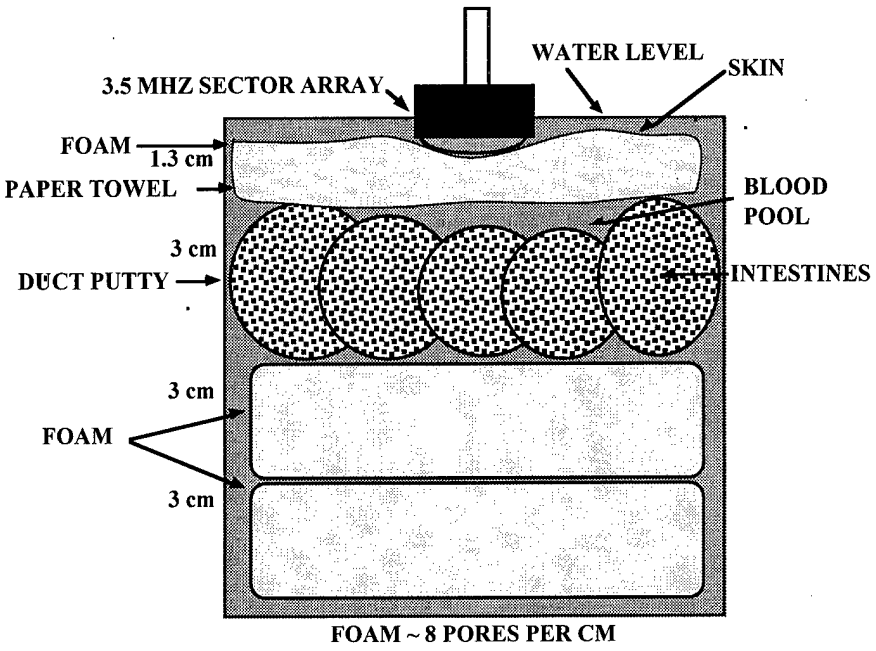
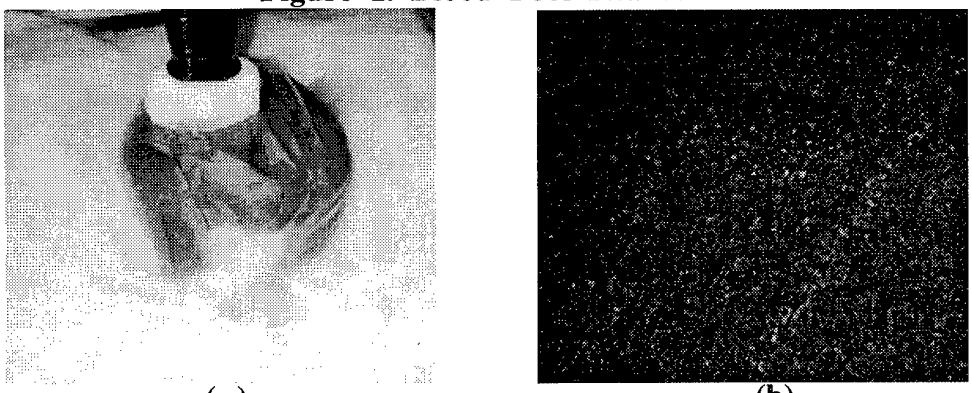


Figure 1. Blood Pool Phantom



(a) **(b)**

Figure 2. Photographs of Duct Putty and Foam Used in the Phantom

Figure 2(a) shows the sector array above the duct putty (in plastic container) and (b) a top view of the black colored foam (~ 8 pores per cm) that was used in constructing the blood pool phantom as shown in figure 1. The foam is very coarse and the large pores can be seen in the photograph.

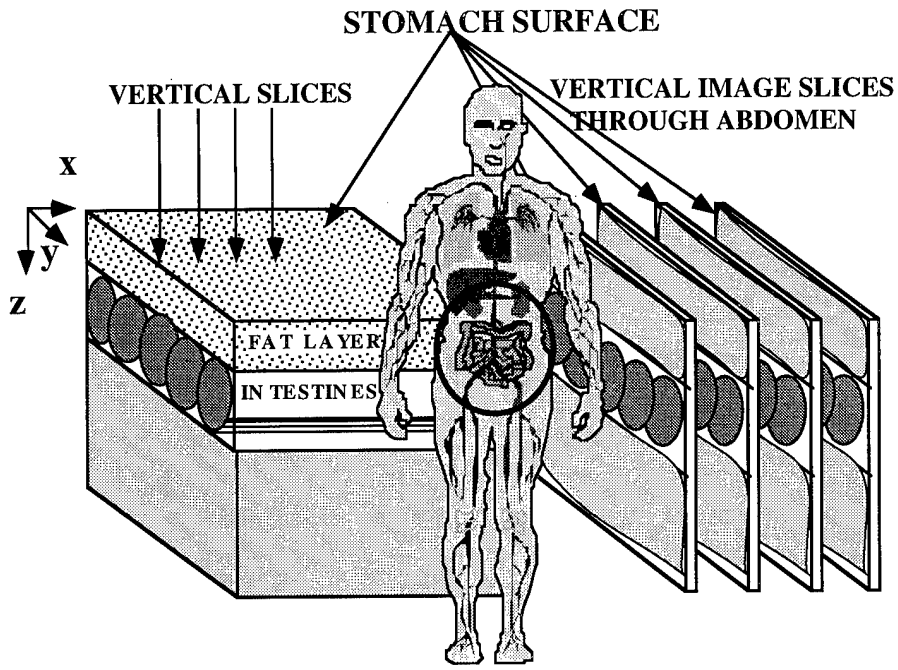


Figure 3. Blood Pool Phantom Slice Image Geometry

Figure 3 is a simple graphical illustration of the image display used in constructing vertical slice images (B-Scans) of blood pool phantoms and humans. The images are as if one cut vertical slices through the phantom (human) and then viewed them as a side view (YZ or XZ planes).

Figure 4 is a B-Scan image of the phantom with a single blood pool pocket as shown by the dark area between the intestines and the upper layer. It is easily seen with the high frequency sector array at 3.5 MHz. The intestines appear as the curved bright lines and entrapped gas in the bowels reflects the ultrasound leaving the dark area below the intestines void of sound waves.

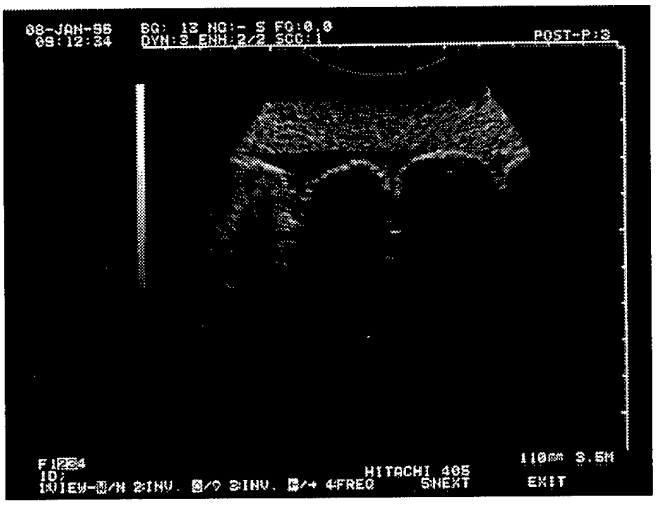


Figure 4. B-Scan Image of Blood Pool Phantom (3.5 MHz)

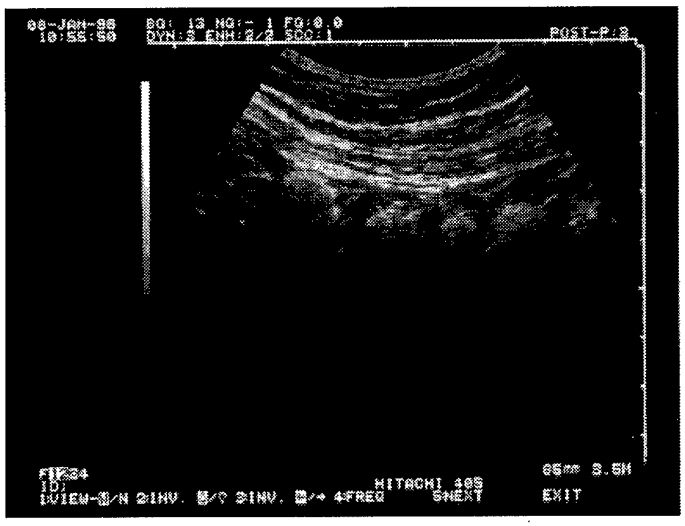


Figure 5. B-Scan Image of Normal (200 LB) Human (3.5 MHz)

Figure 5 is a B-Scan image of the stomach area of a normal (200 lbs) male for comparison with the blood pool phantom image. The dark area or pocket between the intestines and upper layer is absent (no blood pooling). The phantom image appears to be a relatively good model (scattering, etc.) with the insitu image at 3.5 MHz. The actual intestines appear curved in the human as simulated with the duct putty in a plastic container with entrapped air. The layers of muscle between the stomach and the intestines exhibit bright broken lines in the human and uniform scattering in the phantom. These lines can be simulated in the phantom by the insertion of thin paper towels with entrapped air. The phantom was altered and the results were very promising showing similar bright broken lines in the muscle.

There are a few dark areas above the intestines in the normal (200 LB) human (Figure 5), but they appear to have scattering centers within them indicating they are not filled with non-echo-genic blood.

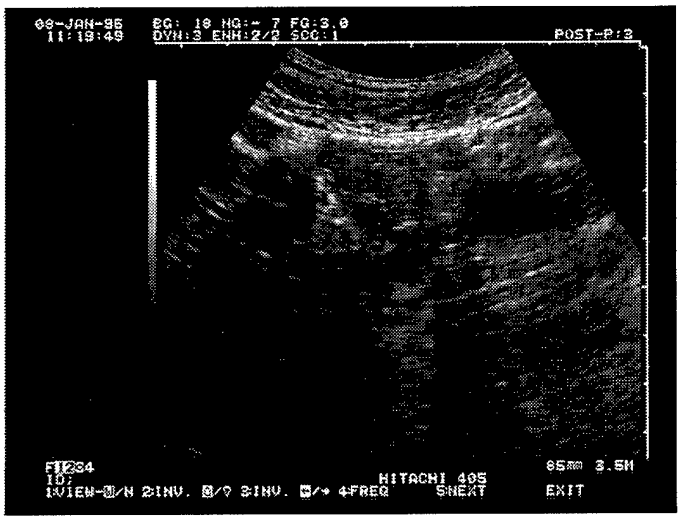


Figure 6. B-Scan Image of Normal (150 LB) Human (3.5 Mhz

Figure 6 is the B-Scan image of the stomach area of a normal (150 LB) human in good physical condition (very low fat content). The bright lines below the surface of the

stomach are more concentrated than the 200 LB human and the area above the intestines has more scattering. The thickness of the fat or muscle layer is thinner on the 150 LB human indicating less fat within the layer.

Conclusion of Phantom Blood Pool Imaging (3.5 MHz) using a Commercial B-Scan with a Sector Array

Blood pool detection and imaging at 3.5 MHz is viable assuming the phantom is representative of a human in this condition. Information from the trauma medical community supports this conclusion and indicates that it is relatively easy to detect blood pooling using ultrasound above 2 MHz. Information of blood pooling detection and imaging using diagnostic ultrasound below 2 MHz appears to be unavailable. We believe the lack of data around 1 MHz is that no commercial B-Scans operate at frequencies below 2 MHz.

If the Army's combat portable "3-D" holographic blood pooling system is to be constructed at 1 MHz, then experimental imaging data should support the decision to reduce the "2-D" array's complexity and cost while retaining sufficient lateral resolution for imaging. The 1 MHz array frequency percent bandwidth should remain approximately the same as the higher frequency systems to insure adequate depth resolution.

Blood pool imaging data must then be collected using a laboratory 1 MHz imaging system to verify that sufficient scattering from the stomach layer muscles and upper surfaces of the intestines will provide acceptable resolution for blood pool imaging.

Single Channel 1 MHz Holographic "1-D" Imaging of Blood Pool Phantoms

A laboratory single channel rectilinear scanned 1 MHz holographic imaging system was configured to image blood pool phantoms. The tests were implemented to verify or refute the viability of using 1 MHz ultrasound in the combat field system employing a small "2-D" array.

The blood pool phantom used in the 3.5 MHz tests with the commercial portable B-scanner was set up in the laboratory water tank for single channel 1 MHz imaging using a focused transducer and the x-y scanner. Figure 7 is a simple cross sectional diagram of the foam duct putty blood pool phantom with a focused transducer above it in the water tank. The blood pool is simulated by water pocket between the upper layer foam (muscle layer) and the duct putty (intestines). The water pocket is non-echo genic and appears in the ultrasound image as a dark area.

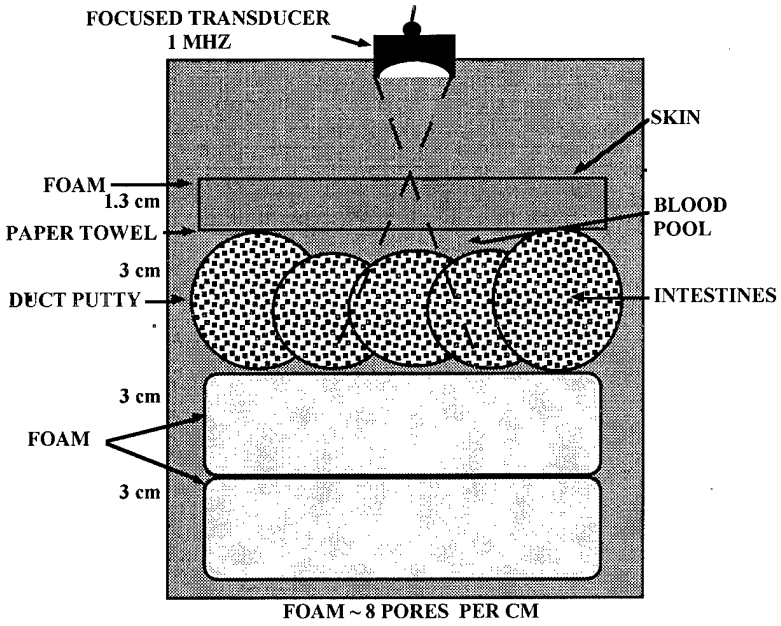


Figure 7. Blood Pool Phantom in water tank for 1 MHz Imaging Tests

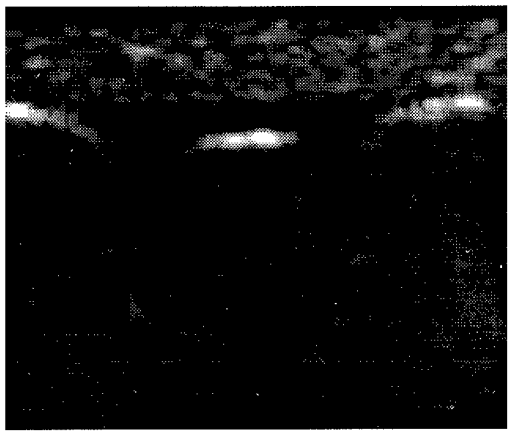


Figure 8. Holographic "1-D" Line Image of Blood Pool Phantom (1 MHz)
Figure 8 is a holographic (B-Scan) image (~ 7.5 cm in length and depth) of the phantom with a blood pool pocket as shown by the dark area between the intestines and the upper layer. The sample density is 0.5mm in the horizontal and 0.3mm in the vertical. It is easily seen at 1 MHz using the scanned holographic system using the focused transducer . The intestines appear as the bright areas (high reflectivity) as a result of entrapped gas in the bowels. The low frequency appears to penetrate around the air entrapped areas of the intestines and reflect from the lower foam layers. The high frequency B-Scan images show dark areas below the intestines indicating loss of ultrasound as a result of higher attenuation at these frequencies.

Single Channel 1 MHz Holographic "1-D" Imaging of Phantom (Normal)

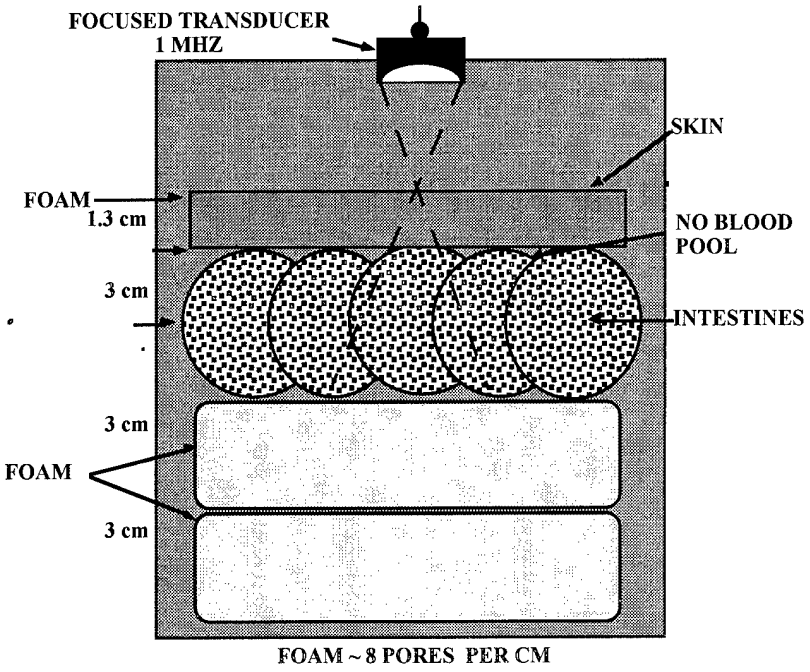


Figure 9. Phantom (No Blood Pool) in water tank for 1 MHz Imaging Tests

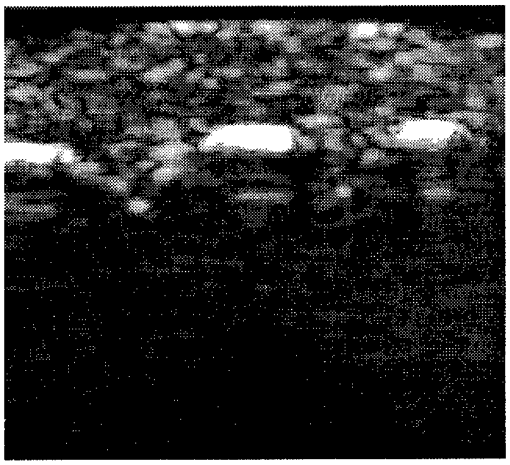


Figure 10. Holographic "1-D" Images of No Blood Pool Phantom (1 MHz)

Figure 9 is the geometry of the ultrasound phantom without blood pooling used in 1 MHz holographic imaging experiments. The phantom is essentially identical to the one used that simulated blood pooling, but without the water cavity between the upper stomach layer and the intestines.

Figure 10 is a holographic (B-Scan) of the no-blood pool phantom (7.5cm length and depth). The sample density is 0.5mm in the horizontal and 0.3mm in the vertical. The 1 MHz image shows the absence of very dark areas (compare with figures 7 & 8) between the stomach muscle layer (upper layer) and the intestines indicating no trauma from internal bleeding in this area. The dark areas below the intestines indicate that the ultrasonic energy is highly attenuated as a result of entrapped air in the intestines. Dark

areas in the images either indicate non-echo genic layers or loss of ultrasound from highly attenuative layers above these regions. To interpret the images correctly one must know where the source position of the ultrasound beam with respect to the x,y,z coordinates.

Comparison of Image Resolution Between Commercial B-Scan (3.5 MHz) and Holographic (1 MHz)

Figure 11 is a composite collection of commercial B-Scan (Hitachi 3.5 MHz) and holographic images (1 MHz) of a blood pool phantom and human for comparing lateral and depth resolutions. Figure 11(a) is a B-Scan image of the blood pool phantom with the dark area (blood pool) between the upper fat or muscle layer and the intestines (curved lines). Figures 11(b)is the 1 MHz holographic image of the blood pool phantom. The lateral resolution at 1 MHz is obviously less than the three focal zone B-Scanner at 3.5 MHz. The dark area (blood pool) in the holographic image (11b) is easily identified and imaged at 1 MHz.

Figure 11(c) is the B-Scan image (3.5 MHz) of a 200 LB normal human (stomach area) showing bright lines interlaced in the muscle layer and the characteristic scallop lines (intestines) below them.

Figure 11(d) is the 1 MHz holographic image of a phantom without blood pooling. The dark area between the stomach layer and intestines is absent indicating no trauma from blood pooling. We assume that a combat wound (internal bleeding) in this area would give rise to significant bleeding in this cavity, and would be easily identified as compared withthe small cavities that were simulated in our phantom.

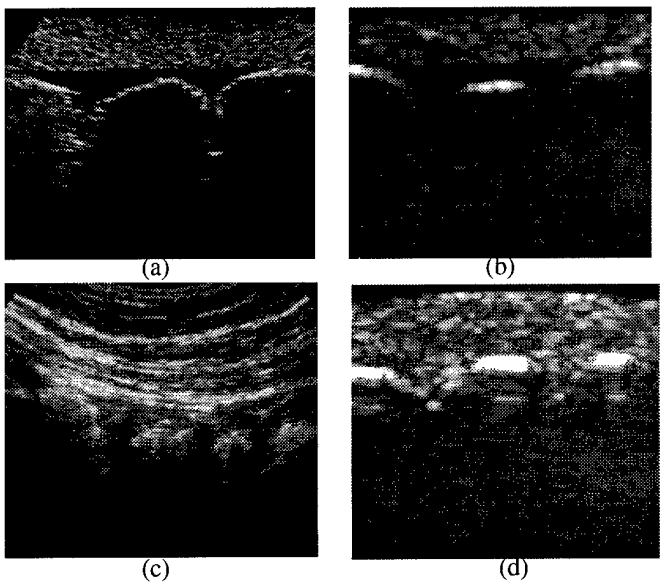


Figure 11. Comparison of B-Scan (3.5 MHz) and Holographic (1 MHz) Images of Blood Pool Phantoms and Human

3D Volumetric Visualization

Figures 12 and 13 illustrate the fully three-dimensional nature of the holographic image datasets. Both figures show a 64x64x256 image (256 depth planes), obtained by a 1 MHz holographic scan, of a blood pool phantom consisting of 3 parallel sections of "intestine".

Figure 12 shows a full-screen view of the user interface provided by the InViVo software package [1]. A volumetric view of the entire data set, viewed obliquely, is shown in the upper left quadrant, with a wireframe representation of the viewing angle shown in the upper right inset. The other three quadrants (outlined in red) show 2-D views of the holographic dataset, resliced on the YZ, XZ, and XY orthogonal axes. (2-D slices at arbitrary angles can also be obtained, but this capability is not illustrated here.) Figure 13 shows full-size volumetric images of the dataset from two angles.

At present, it appears that the major value of volumetric visualization for blood pool detection is to provide a 3-D context within which to evaluate 2-D slice images. However, the ability to reconstruct (select) arbitrary 3-D slices from the holographic datasets should provide a powerful tool for interpretation.

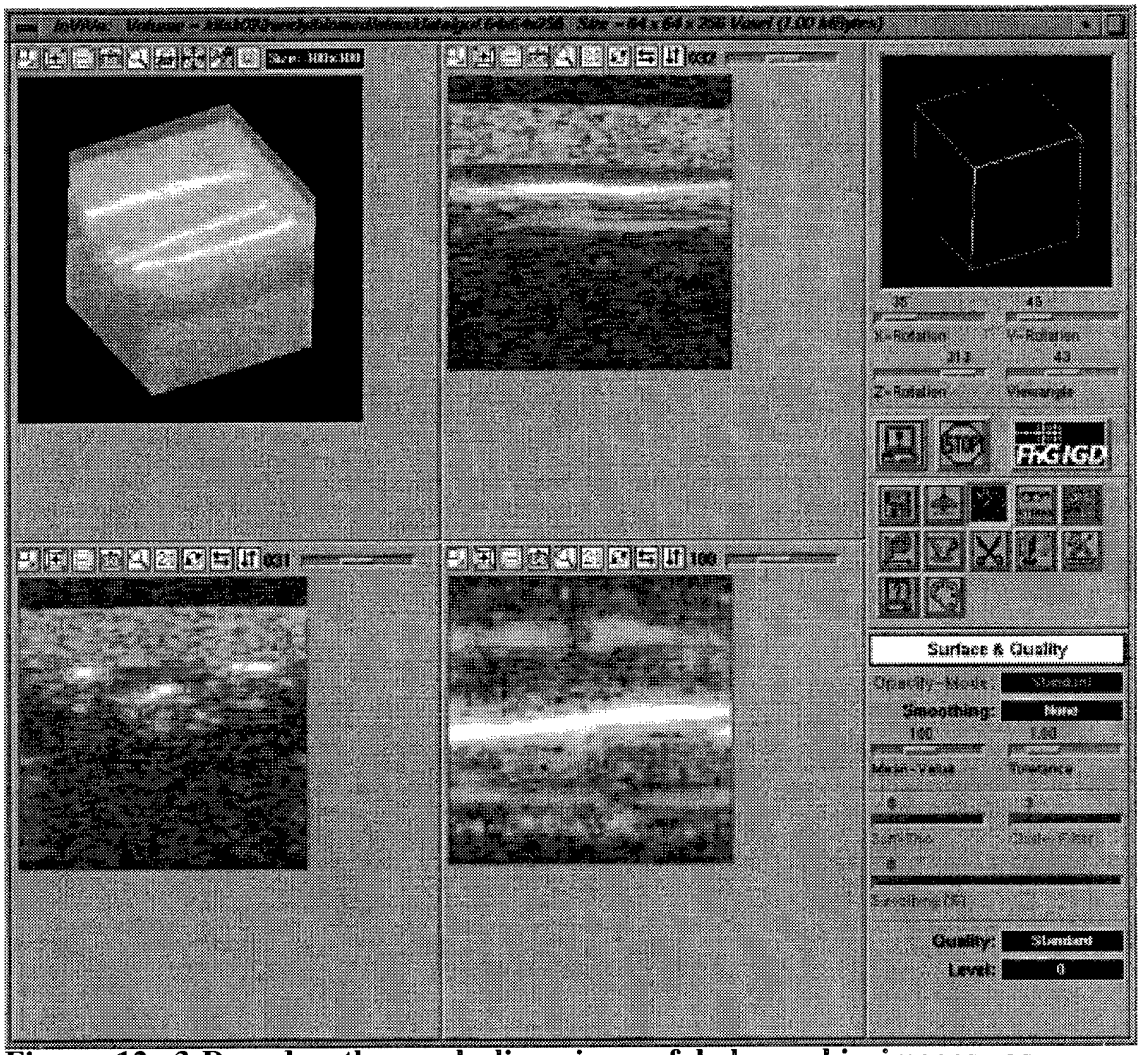


Figure 12. 3-D and orthogonal slice views of holographic images, as rendered by InViVo software package (full screen).

[1] InViVo is a product of the Fraunhofer Center for Research in Computer Graphics, Providence, RI.

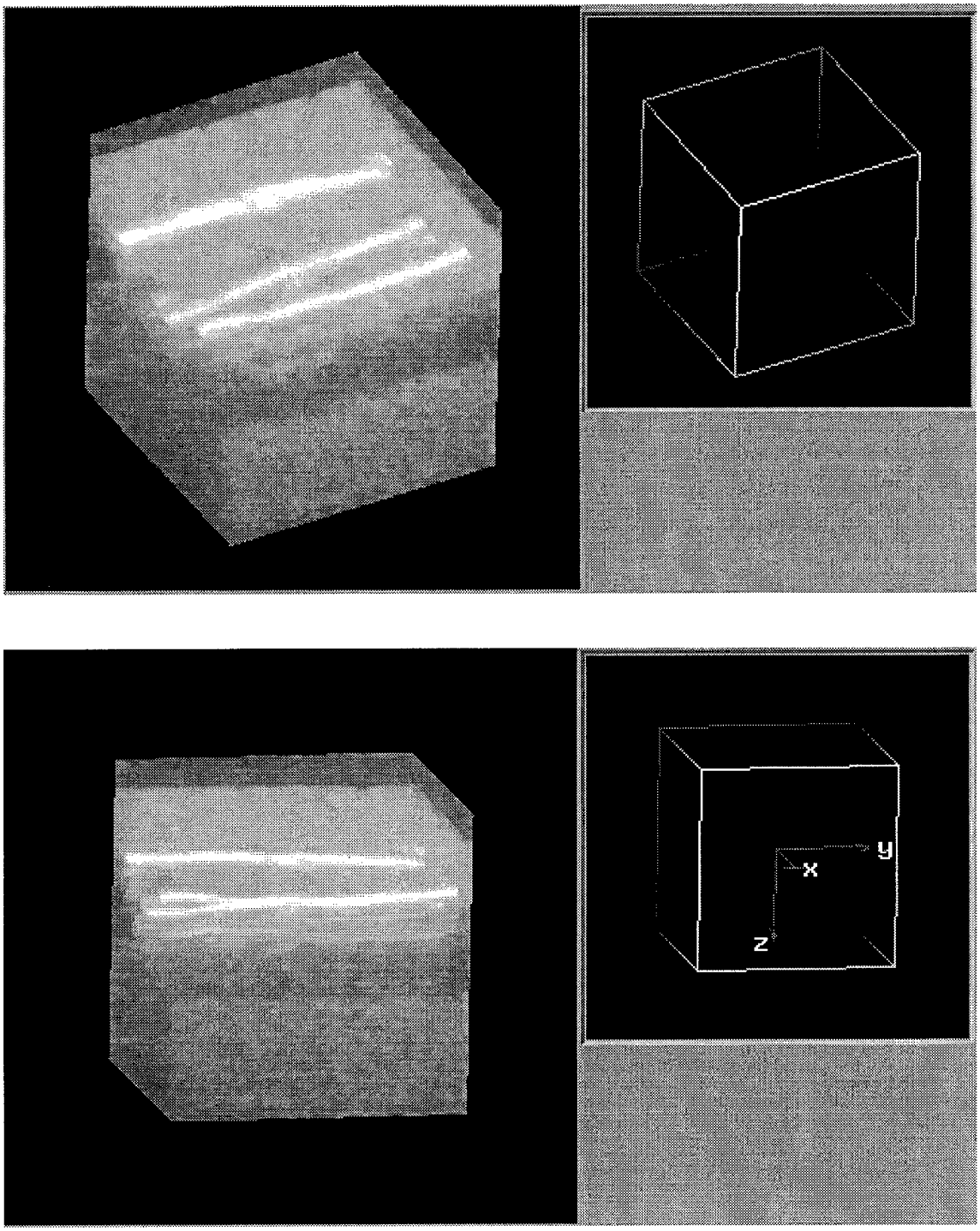


Figure 13. Volumetric views (Maximum Intensity Projection) of holographic images of the blood pool phantom.

Conclusion of Phantom Blood Pool Imaging (1 MHz) using PNNL's Holographic Imaging System

Blood pool detection and imaging at 1 MHz appears to be viable assuming the phantom is representative of a human in this condition. University of Texas medical staff working in the field of trauma gave PNNL directions for constructing a phantom similar to those they use in B-Scan training for blood pooling. On the assumption that the phantom is representative of the human body at 1 MHz, blood pooling is detectable and can be imaged at this frequency.

If the Army's combat portable "3-D" holographic blood pooling system is to be constructed at 1 MHz, sufficient experimental imaging data at 1 MHz should support this important decision. The next phase of this program (building the "2-D" array) depends on the successful results of 1 MHz blood pool imaging with the "1-D" array. The "2-D" array's complexity and cost would be significantly reduced if 1 MHz can be used.

Appendix 3:

"Coherent Lens-based Image Reconstruction"

Coherent Lens-based Image Reconstruction

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March 4, 1996

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Abstract

This paper demonstrates how holographic, or Fourier optic, image reconstruction techniques may be applied to a lens-based coherent imaging system. These techniques overcome the limitations of lens-based systems such as limited resolution and very limited depth-of-field (or depth-of-focus). Additionally, these techniques are suitable for full three-dimensional high-resolution imaging.

1.0 Introduction

Efforts are presently under way to fabricate large-scale two-dimensional arrays for ultrasonic imaging applications. For example, one such system is being designed by Loral and will consist of a 2-D array of 128 by 128 elements. The element spacing is 200 μm, or two-thirds wavelength at the center frequency of 5 MHz Thus, the array will be approximately 25.6 mm square. Each element will allow quadrature detection (I and Q) of the scattered wavefront. The present design will use coherent external illumination sources. A lens (7 - 10 cm diameter) will be used to increase the image size (demagnify) so that the visible aperture is on the order of 7 - 10 cm. An additional advantage of the large lens is that it will capture the ultrasonic energy over a larger aperture than would imaging with the array without using a lens.

Efficient high-resolution imaging is possible using a 2-D array without using a lens at all. The sampled wavefront can be 'back-propagated,' using holographic (Fourier optic [1]) image reconstruction techniques to form the image that existed at the object plane. This image reconstruction technique utilizes the 2-D Fourier Transform to decompose the scattered wavefront into a number of discrete plane wave components. These plane wave components can then be phase-shifted (or back-propagated) to the object plane. The inverse Fourier Transform then yields the focused image. A larger effective image size can be obtained by artificially increasing the size of the data array by zero-padding. This type of image reconstruction has also been extended (at PNNL) to true three-dimensional imaging by utilizing coherent wide-bandwidth illumination. The I and Q waveforms can be digitized over a time interval and the time series can then be Fourier Transformed to yield the scattered wavefronts magnitude and phase over a wide frequency bandwidth. This data can then be back-propagated resulting in a fully focused 3-D image of the object.

Advantages of this type of synthetic processing are that no paraxial, Fresnel, or Fraunhofer approximations are made. Therefore, the target objects can be close to the array, and can fill the array aperture without significant degradation of the theoretical diffraction limited resolution. This resolution is approximately

$$\delta_{\text{lateral}} = \frac{\lambda}{2}\left(\frac{R}{L}\right) \tag{1}$$

$$\delta_{\text{depth}} = \frac{u}{2B} \tag{2}$$

where λ is the wavelength, R is the range to the target, L is the aperture size, u is the speed of wave propagation, and B is the bandwidth. Generally it is possible to achieve F-1 or F-2 for resolution of one-half to one wavelength, where F-n refers to the optical F-number or *R/L*.

However, there may be advantages in using a lens in this type of imaging system. The aperture over which the acoustic energy is gathered can be significantly larger than the array. This may lead to an enhanced signal-to-noise ratio. Also, a lens may be used effectively to magnify or demagnify the image size. The techniques shown in this paper may be used to essentially *'fine-tune'* the images obtained from a lens-based imaging system to obtain performance near the theoretical limits.

2.0 Lens-based Image Reconstruction

Data is gathered from a coherent lens-based imaging system, as depicted in Figure 1. Since the system is coherent, the magnitude and phase of the complex wavefront that strikes the lens are measured by the imaging system. If the lens is properly designed, and the array is placed at the correct focal zone, then the image of the target object will be obtained simply by taking the magnitude of the measured signal, $s(x,y) = \sqrt{I^2 + Q^2}$. This image will appear inverted due to the image-reversing characteristic of the lens. If the lens can be assumed to be 'thin,' then the location of the array will obey the Gaussian Lens Formula for optimal focus

$$\frac{1}{s_o} + \frac{1}{s_i} = \frac{1}{f} \tag{3}$$

where f is the focal length of the lens.

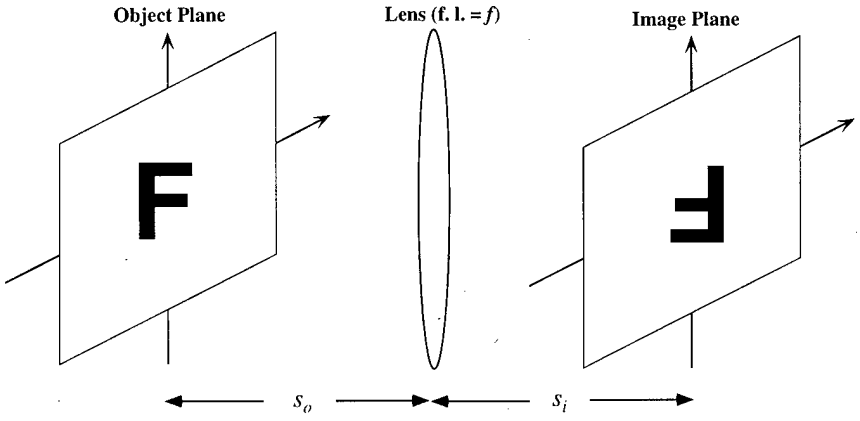

Figure 1 Lens-based imaging system.

2.1 Thin-lens requirement

All of the analysis done in this section relies on the thin-lens approximation. The critical element of the thin lens requirement is that a ray entering the lens at coordinates (x,y) can be approximated as leaving the lens at the same (x,y) coordinates. This allows wave propagation through the lens to be approximated as a simple phase-shift of the wave.

2.2 Lens-based imaging simulation

To simulate a lens-based imaging system, the scattered wavefront from a test target composed of 12 points arranged to form the letter F was computed by superposing the complex phase of each of the point scatterers over an (x,y) aperture. This complex phase is simply $e^{-jk\sqrt{(x-x_s)^2+(y-y_s)^2+z_s^2}}$ where $k = 2\pi/\lambda$, $(x,y,0)$ are the aperture coordinates, and (x_s,y_s,z_s) are the source point coordinates. This aperture was chosen to be just on the object side of the lens plane. The real-part of this scattered wavefront is shown in Figure 2. The imaging system was configured for $s_o = s_i = 2f$.

The wavefront (of Figure 2) was passed through the lens by multiplying by $e^{-j\phi(x,y)}$ where $\phi(x,y)$ is the phase-shift through the lens. This phase function was chosen so that a plane-wave normally incident on the lens aperture would focus at the focal

length, *f*. The real-part of the phase-front after passing through the lens is shown in Figure 3.

After passing through the lens, the phase-front will converge as it propagates toward the array. This propagation can be determined using Fourier optics techniques[1]. The phase-front can be 2-D Fourier Transformed which decomposes the phase-front into discrete plane wave components. These plane wave components can then simply be phase-shifted to determine their amplitude and phase at any other depth, *z*. The actual wavefront at this depth is then determined by inverse 2-D Fourier Transform of the plane-wave components. The magnitude of the wavefront at the focal-plane is shown in Figure 4. The image is inverted (as expected) and is reasonably well focused with each discrete point in the F clearly resolved. There is, however, some undesirable spreading of each point's response. This is perhaps due to the non-paraxial nature of the example simulated.

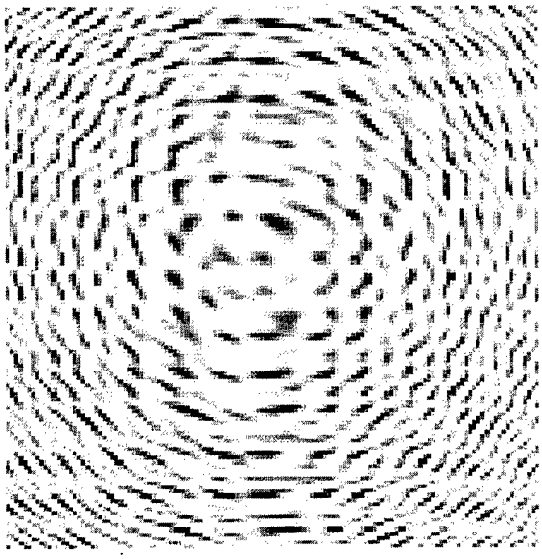

Figure 2 Object phase-front just before impinging on the lens.

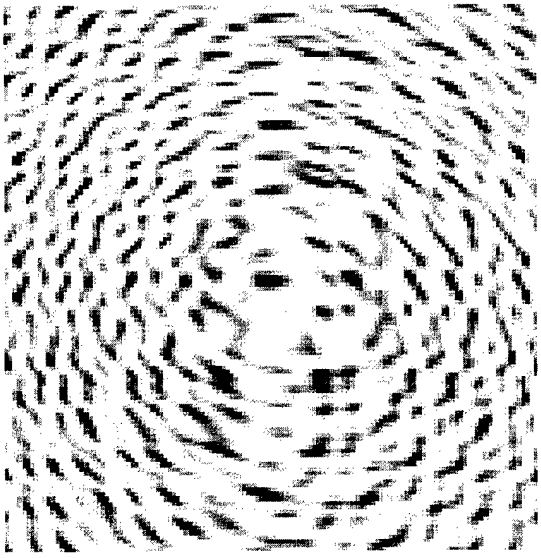

Figure 3 Object phase-front just after passing through the lens.

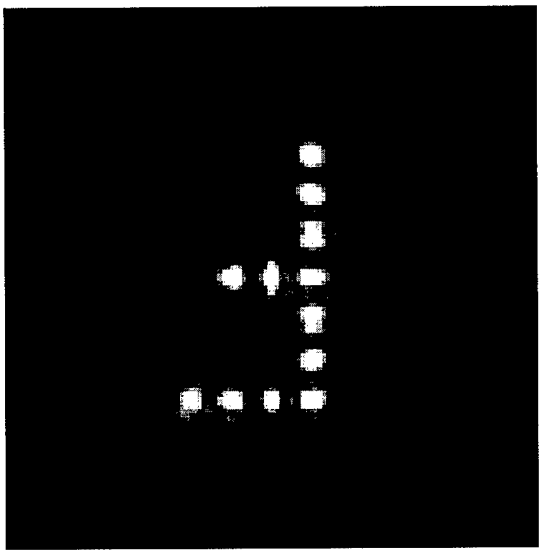

Figure 4 Image magnitude at the focal plane.

2.3 Correction of non-ideal array placement

Fourier optics techniques can be used to correct for placement of the array at positions other than the ideal focal plane. Figure 5 shows an image that would be obtained if the array were placed at 75% of the correct distance from the lens. The coherent data obtained by the array, can however, be 'forward-propagated' using

plane-wave decomposition to the correct focal plane of the array. This results in the image shown in Figure 6, which is essentially identical to the image obtained in Figure 4.

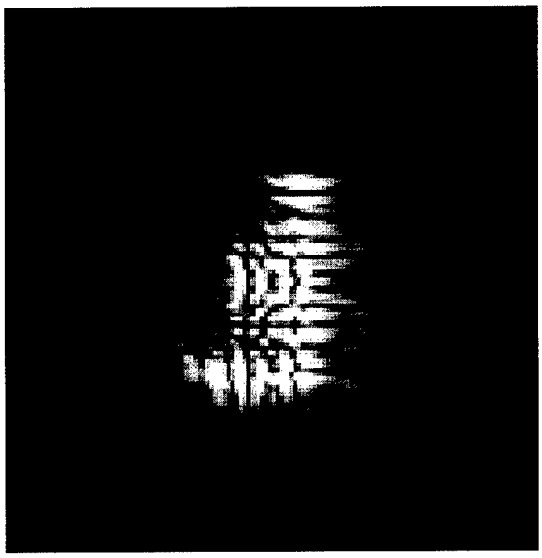

Figure 5 Image magnitude at 75% of the focal distance.

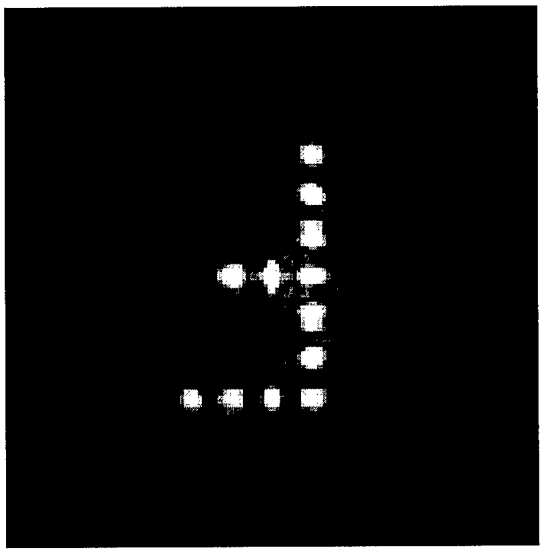

Figure 6 Image magnitude after correction to the correct focal distance using 'forward-propagation.'

2.4 Back-propagation to the object plane

A superior image formation method to that discussed in Section 2.3 is available. Instead of simply forward or back propagating the wavefront to the focal plane, the wavefront can be back-propagated to the lens plane, phase-shifted through the lens, and back-propagated to the object plane. This forms a virtual image of the object. Most importantly, this technique removes the effect of the lens. Therefore, as long as the shape of the lens is known, any aberration that it causes in the image can be removed by this process. The reconstructed image of the F test object obtained in this manner is shown in Figure 7. This image is tightly focused and does not have the sidelobe artifacts of the lens-only reconstruction shown in Figures 4 and 6. Note that in Figure 7 the image-reversal is not present.

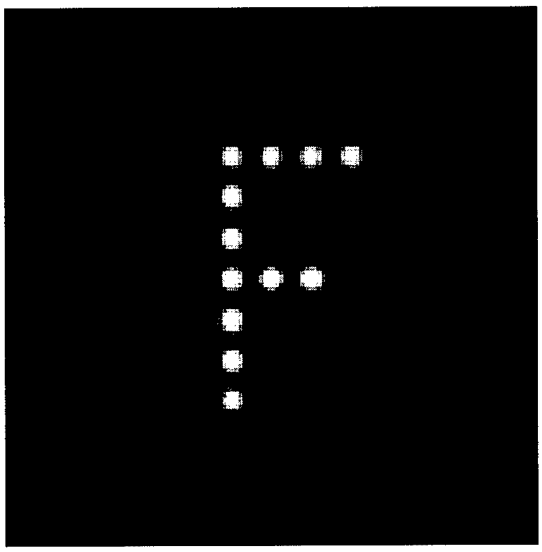

Figure 7 Image magnitude after 'back-propagation' with phase-correction for the lens.

2.5 Application to full 3-D imaging

The example demonstrated in Figures 2-7 was simulated using a single frequency. These techniques are, however, fully adaptable to 3-D imaging. The effect of the lens on each frequency component of a wideband system can be eliminated using the technique demonstrated in Section 2.4. After the effect of the lens is removed, the efficient 3-D holographic reconstruction techniques (developed at PNNL) can be

applied. Three-dimensional imaging is important in the lens-based imaging system, since it will allow a virtually unlimited depth-of-field (or depth-of-focus).

Conclusions

Holographic, or Fourier optic, reconstruction techniques can be applied to lens-based imaging systems. The only significant limitation in the analysis is that the thin-lens approximation must be valid. These techniques have been demonstrated on simulated single-frequency data. The extension to full three-dimensional imaging is straightforward. Three-dimensional imaging using reconstruction techniques developed at PNNL would allow fully-focused three-dimensional imaging with an unlimited depth of field using a 2-D array, with or without a lens. If a lens is used, known aberrations that it possesses may be removed from the images algorithmically.

References

[1] Goodman, J. W., **Introduction to Fourier Optics**, McGraw-Hill, 1968.